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Responses to gait perturbations in stroke survivors who prospectively experienced falls or no falls

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ABSTRACT

Background: Steady-state gait characteristics appear promising as predictors of falls in stroke survivors. However, assessing how stroke survivors respond to actual gait perturbations may result in better fall predictions. We hypothesize that stroke survivors who fall have a diminished ability to adequately adjust gait characteristics after gait is perturbed. This study explored whether gait characteristics of perturbed gait differ between fallers and non fallers. *Method:* Chronic stroke survivors were recruited by clinical therapy practices. Prospective falls were monitored over a six months follow up period. We used the Gait Real-time Analysis Interactive Lab (GRAIL, Motekforce Link B.V., Amsterdam) to assess gait. First we assessed gait characteristics during steady-state gait and second we examined gait responses after six types of gait perturbations. We assessed base of support gait characteristics and margins of stability in the forward and medio-lateral direction. *Findings:* Thirty eight stroke survivors complete our gait protocol. Fifteen stroke survivors experienced falls. All six gait perturbations resulted in a significant gait deviation. Forward stability was reduced in the fall group during the second step after a ipsilateral perturbation. *Interpretation:* Although stability was different between groups during a ipsilateral perturbation, it was caused by a secondary strategy to keep up with the belt speed, therefore, contrary to our hypothesis fallers group of stroke survivors have a preserved ability to cope with external gait perturbations as compared to non fallers. Yet, our sample size was limited and thereby, perhaps minor group differences were not revealed in the present study.

1. Introduction

Fall rates are high in the chronic stage after stroke (Weerdesteyn et al., 2008) and higher than in healthy older adults (Weerdesteyn et al., 2008). Most falls occur during gait (Forster and Young, 1995) and consequently assessment of gait could be useful in predicting fall risk. Assessing quality of steady-state gait may quantify how the system handles small, internal perturbations like neuromuscular noise (Bruijn et al., 2013; Dingwell et al., 2000). Interestingly, stroke survivors have a more variable gait pattern and a reduced quality of gait as compared to healthy controls (Kao et al., 2014; Punt et al., 2016). Moreover, quality of gait shows promise as a predictor of falls in stroke survivors (Mansfield et al., 2015b; Punt et al., 2016).

Other aspects than the quality of steady-state gait might contribute to the prediction of fall risks in stroke as well. Large, external gait perturbations experienced in everyday life, like trips and slips, may require a substantial change of the gait pattern to overcome the perturbation and prevent a fall (Kajrolkar et al., 2014; Kajrolkar and Bhatt, 2016; Krasovsky et al., 2013). Thus, measures of how subjects react to larger perturbations are interesting in relation to fall prevention. Stroke survivors appear to respond less effectively to external gait perturbations (Krasovsky et al., 2013). Thus external gait perturbations may provide additive information with respect to fall risk in stroke survivors.

It is currently unknown if, and how, gait recovery characteristics, after a gait perturbation are associated with falls in stroke sur-

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vivors. This study attempts to discover the potential of using gait perturbations to predict falling in stroke survivors. Therefore, our aim was to explore whether differences exist in responses to external gait perturbations between a group of stroke survivors that experienced a fall in daily life, and a group that did not.

We focused on gait recovery characteristics that reflect how and to what extent stroke survivors are able to cope with external gait perturbations. Perturbations of gait require adequate base of support (BoS) adjustments through adapting foot placement. Dynamic stability quantified by the margins of stability (MoS) (Hof, 2008; Hof et al., 2005) provides additional information by relating the kinematic state of the body center of mass (CoM) to the BoS. We prospectively studied the relation between gait adaptations after a perturbation and fall risk. We hypothesized that stroke survivors who fall during follow-up have less effective adaptations of foot placement after gait perturbations coinciding with smaller MoS than stroke survivors who do not fall during follow-up.

2. Method

We recruited stroke survivors through flyers in physical therapy practices and various national peer group meetings in the Netherlands. Stroke survivors were recruited if they were at least six months post-stroke, aged at least eighteen and lived independently in the community. We excluded stroke survivors with a functional ambulation category lower than 3 (Holden et al., 1984), a minimal mental state examination (MMSE) lower than 25 (Folstein et al., 1975) and or other disorders such as neurologic, musculoskeletal, respiratory or severe cardiovascular disorders that affected gait performance. The medical ethics committee 'Noord Brabant, The Netherlands' approved the research protocol and treatment of the participants was according to good clinical practice. Prior to the gait analysis, demographic and stroke specific characteristics were collected such as; sex, age, body length and weight, time since stroke, hemiparetic side, use of a walking aid, use of medication.

2.1. Experimental set up

All participants walked on the Gait Real-time Analysis Interactive Lab (GRAIL, Motekforce Link B.V., The Netherlands). The GRAIL consists of: a motion-capture system (Vicon, Vicon Motion Systems, UK) with ten infrared cameras (Bonita B10, Vicon Motion Systems, UK), a dual-belt treadmill with two embedded force platforms and synchronized virtual environment (Motekforce Link B.V. The Netherlands). A custom written application in D-flow software (Motekforce Link B.V. The Netherlands) controlled the GRAIL.

Participants wore tight fitting black clothes. In order to collect full body kinematics we used a the human body model based on 47 passive markers (van den Bogert et al., 2013) These were placed before the gait analysis by the same investigator throughout the study to maximize consistency between participants. Furthermore participants wore a safety harness which prevented actual falls.

2.2. Gait protocol

Twenty-four hours prior to clinical and laboratory testing participants were asked not to drink any alcoholic beverages and to avoid any other activities that could affect physical performances. All measurements were performed during a single visit at the rehabilitation center Revant, Breda, The Netherlands. After participants became familiarized to walking on the treadmill, we first assessed steady-state gait characteristics during sixty consecutive strides at a gait speed of 0.41 m/s. Subsequently, all perturbations were executed at the same gait speed of 0.41 m/s. In pilot experiments, this gait speed in combination with perturbations was found to be feasible for most community walking stroke survivors.

The perturbation protocol consisted of two separate trials; each trial comprised 16 perturbations; each perturbation was followed by a wash-out period of on average 15 s. Perturbations were triggered by foot contact (FC). The sequence of the perturbations was semi random as the perturbation type was fixed but the triggering at the left or right foot placement was random. Participants were allowed to hold the handrail during the first four perturbations, those perturbations were not included in the analysis. Each trial lasted for four minutes. Between trials breaks were taken to avoid fatigue as much as possible.

The first perturbation trial contained medio-lateral (ML) perturbations. More specifically, the walking surface of the treadmill moved either to the left or right side at FC of the participant (see Fig. 1 for an illustration and Fig. 2, ML Perturbation for the perturbation intensity). Depending on whether right or left FC was followed by a right or left walking surface translation, the perturbations were classified as "ipsilateral" or "contralateral" gait perturbations. From a static perspective we may expect that during ipsilateral perturbations participants respond quickly, because the supporting limb shifts away from the vertical projection of the CoM, (see Fig. 1 ipsilateral perturbation), which requires an immediate response to maintain stability. In contralateral perturbations (see Fig. 1 contralateral perturbation), the supporting limb shifts toward the vertical projection of the CoM, which may not require an immediate response. However, it should be noted that this explanation holds for static situations while gait is a dynamic activity.

The second perturbation trial comprised anterior-posterior (AP) decelerating perturbations. At either right or left FC the belt speed on the side of the FC decelerated toward 0 m/s and subsequently accelerated toward 0.41 m/s (see Fig. 2, AP Perturbation for an illustration).

As a response with either the paretic leg or non-paretic leg could make a substantial difference, we subdivided the two perturbations types into "response non paretic leg" (NPL) and "response paretic leg" (PL). All perturbation types started 80–90 ms after FC was detected. The maximum ML displacement was 0.045 m and the maximum peak deceleration of the belt speed was 3.9 m/s^2, see Fig. 2 for an illustration. To summarize, we explored a total of six different gait perturbations.

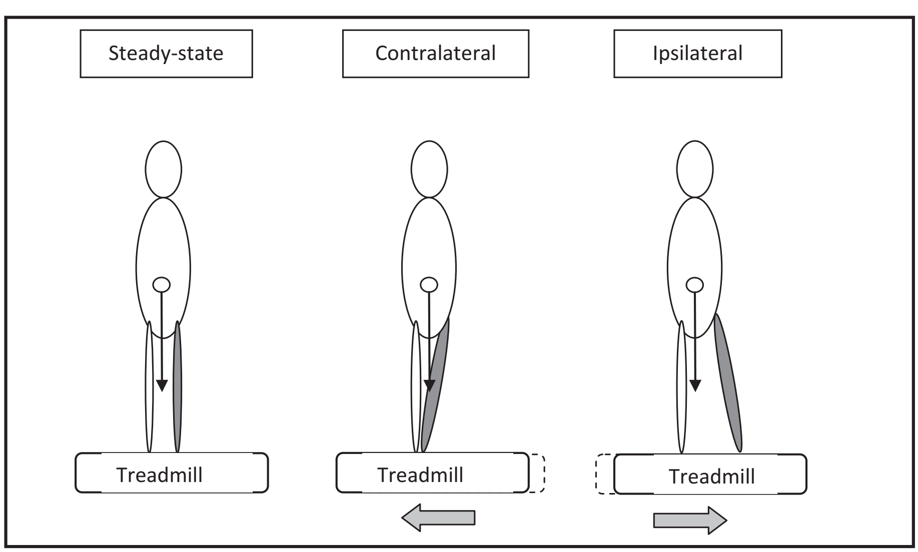

Fig. 1. Backward perspective at right foot contact during medio-lateral treadmill displacements. Left panel represents steady state gait, mid panel represents a contralateral perturbation and the right panel represents an ipsilateral perturbation. Horizontal arrows show the direction of the treadmill displacement. Due to the medio-lateral treadmill displacement in the mid panel, the right foot shifts toward the projected CoM (vertical arrow). In the right panel the right foot shifts away from the projected CoM. The shaded limb represents the limb that was perturbed.

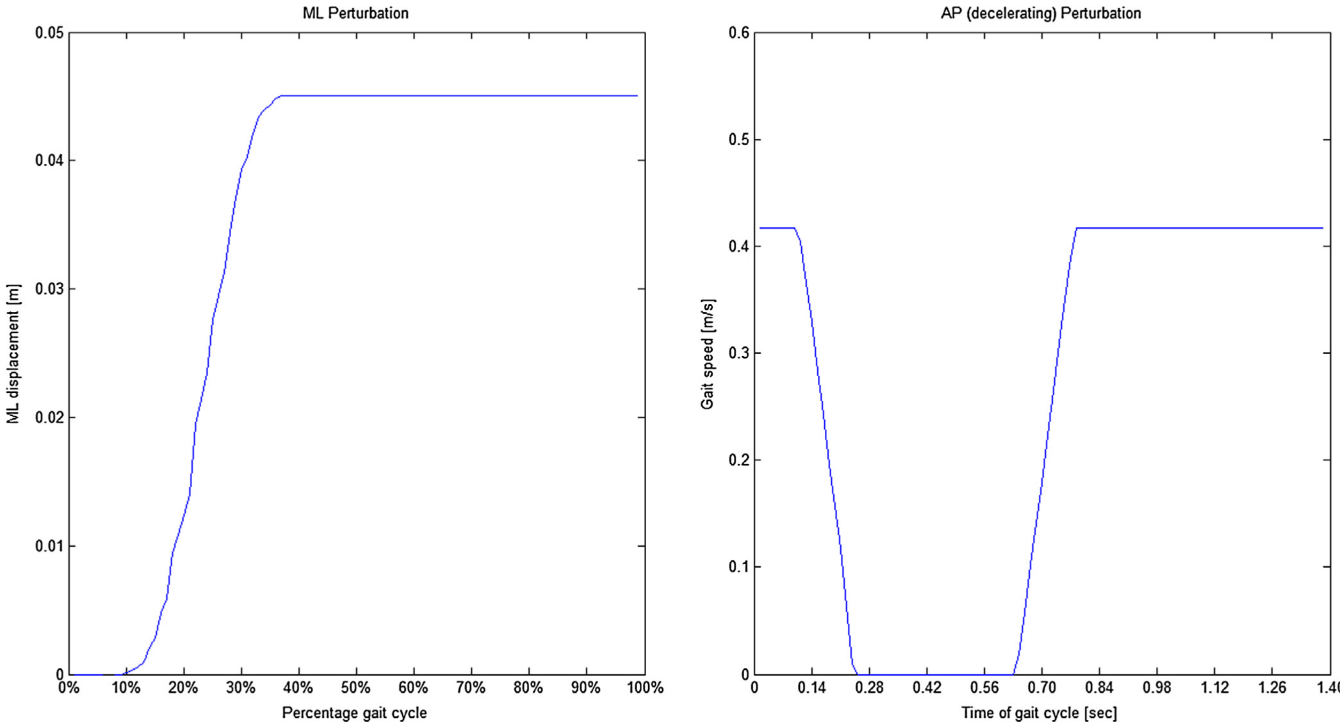

Fig. 2. Gait perturbation in medio-lateral, left panel and anterior-posterior direction, right panel relative to the gait cycle.

Four ML gait perturbations divided into contralateral and ipsilateral and response with either NPL or PL. The final two AP decelerating gait perturbations were divided into "response non paretic leg" (NPL) and "response paretic leg" (PL).

2.3. Data analysis

Discrete gait events like FC were detected using a center of pressure method (Roerdink et al., 2008). Based on these FC events and markers placed at the heel, lateral malleolus and toe on both feet, we calculated step time and the BoS gait characteristics: step length and step width. The whole body CoM was determined using a 14 body segment model (Zatsiorsky, 1998). Subsequently, dynamic stability expressed as the MoS in forward (FW) and ML direction was determined at FC (Hof et al., 2005). A larger MoS indicates a increased dynamic stability. For steady-state gait, the average of these parameters was calculated over 60 strides. The final two perturbations were free of handrail support and were used for further evaluation. Response characteristics were determined at FC of up to six steps after the perturbation. All analyses were performed using custom written Matlab programs (Matlab 2013B).

2.4. Fall status

Falls were detected using a 'fall calendar' and monthly phone calls during six months follow-up. A fall was defined as 'any unanticipated event that results in a participant coming to the ground, floor or lower level' (Lamb et al., 2005). Falls were excluded if the cause was clearly different from a loss of balance, such as when fainting or experiencing an epileptic seizure.

2.5. Statistics

Participants were assigned to the fallers group of stroke survivors if they had experienced at least one fall during follow-up and otherwise in the non fallers group of stroke survivors. Demographic and stroke specific characteristics were compared using an independent samples *t*-test or for not normally distributed variables a Mann Whitney *U* test. Dichotomous variables such as use of a walking aid and sex were examined using a chi square test.

Steady-state gait characteristics were compared between groups using an independent samples *t* test. Next, we examined the perturbed gait characteristics. We first assessed if and how many steps the characteristics after perturbation deviated from state steady gait. We used a dependent samples *t* test to compare each step after the perturbation, with steady-state gait. Results indicated that at least one out of five examined gait characteristics significantly deviated up to six steps after the perturbation (see Appendix A). For further analysis, we therefore included 6 steps. We performed a mixed model ANOVA with steps as our within factor, and fall status as our between subjects factor. The dependent variable was the characteristic of interest. If a main effect of group or interaction effect with group was found, independent samples *t* tests per step were performed to determine in which step (s) groups differed from each other. Similar analysis were performed with preferred steady-state gait speed as covariate, to test for a possible confounding effect, results are shown in Appendix B. A *p*-value of $<.05$ was considered significant; all statistical analysis were performed in SPSS version 23.

3. Results

A total of 38 stroke survivors successfully completed the gait assessments. Fifteen (39%) stroke survivors reported at least one fall. Demographic and stroke specific characteristics did not differ between both groups of stroke survivors, except for the use of a walking aid which was more often used in the fallers group, see also Table 1.

3.1. Steady-state gait

Gait characteristics of the groups were similar during steady-state gait at a fixed speed, except for step time of the paretic leg and step length of the non paretic leg, which were significantly lower in the *F* group, see Appendix A.

3.2. Perturbations

3.2.1. Medio-lateral contralateral perturbations

Overall, contralateral gait perturbations when responding with the non paretic leg (Fig. 3 contralateral NPL) resulted in similar gait characteristics to steady-state gait in the first step, but step length was increased during the second and third step. In addition, step width increased from the second step onwards. MoS ML increased in the first step, (Fig. 4 contralateral NPL, for statistics see Appendix A). No main effects of group or interaction effects with group were found for any of the five gait characteristics, for this perturbation type (Table 2).

Contralateral gait perturbations when responding with the paretic leg (Fig. 3, contralateral PL) showed increased step times

Table 1
Demographic and stroke specific characteristics.

	NF-SS (23) Mean (sd)	F-SS (15) Mean (sd)	*P*-value
Age(y)	**55.0 ± 12.2**	**65.4 ± 6.7**	**.02**
Gender (female/male)	13/10	7/8	.74
Hemiparetic side (right/left)	16/7	10/5	1
Time since stroke (months)	73.8 ± 53	104 ± 89	.25
Number of strokes > 1	3	0	.53
Weight (kg)	87 ± 19	83 ± 20.1	.67
Length (cm)	172 ± 10	171 ± 13	.73
BMI (kg/m^2)	29.5 ± 6.5	28.7 ± 6.1	.78
FAC score	**4.6 ± 1.1**	**4.1 ± 0.9**	**.04**
Use of walking aid (no/yes)	**19/4**	**10/5**	**<.01**
Use of medicines (no/yes)	2/21	2/13	1
MMSE (max 30)	28.3 ± 2.1	27.6 ± 2.0	.41
Preferred gait speed (m/s)	**0.72 ± 0.3**	**0.5 ± 0.28**	**.02**

Mean ± standard deviation from demographic and stroke specific characteristics.
P-values are based on independent sample *t*-test, Mann-Withney *U* test or chi-square tests. Significant differences are printed in bold.

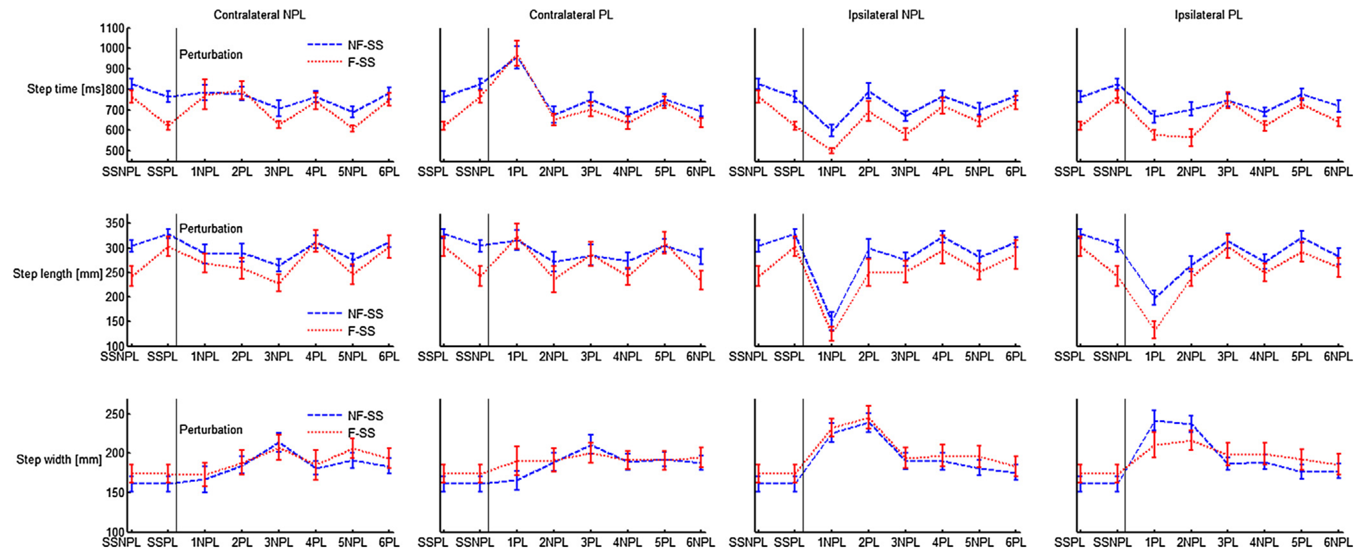

Fig. 3. Step time and base of support (BoS) gait characteristics during steady state (SS) and after gait was medio-lateral perturbed for the paretic leg (PL) and non paretic leg (NPL).

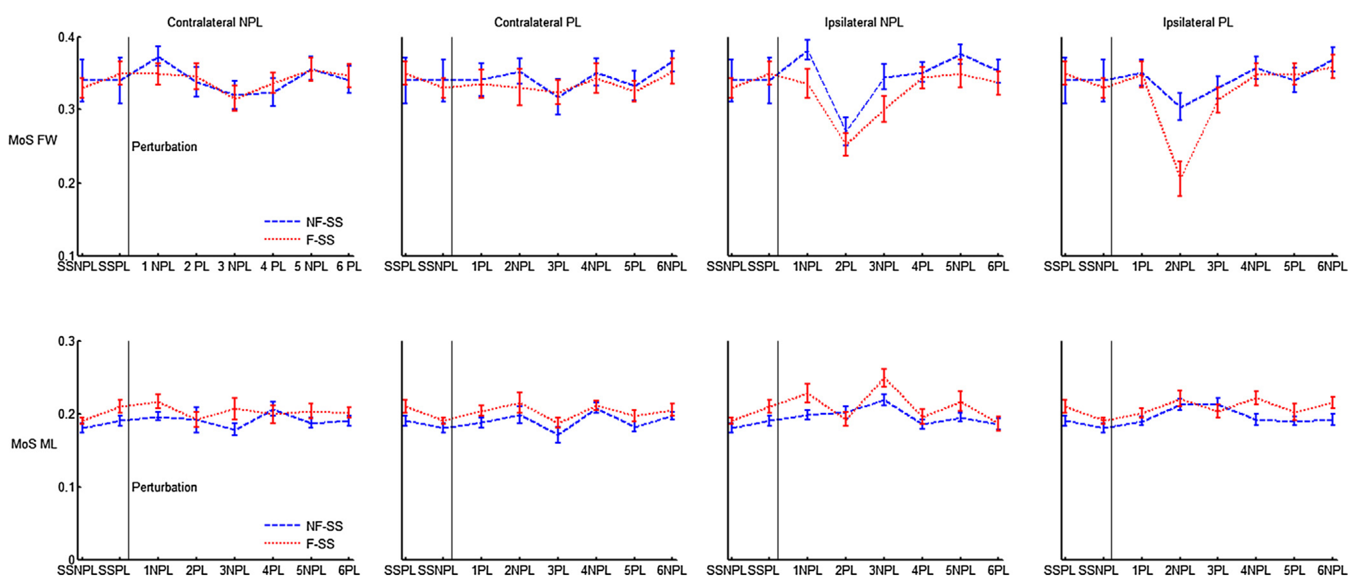

Fig. 4. Margins of Stability (MoS) in the forward (FW) and medio-lateral (ML) direction during steady state (SS) and after gait was medio-lateral perturbed for the paretic leg (PL) and non paretic leg (NPL).

Table 2
Mixed model ANOVA for ML gait perturbations. With the gait characteristic as dependent variable. Number of steps as within factor and group as between effect. Significant group and interaction effects are printed in bold.

Gait characteristic	Effect	*F*	*P*-value
Contralateral perturbation First response Non Paretic Leg			
Step time	Steps	5.21	.01
	Group	0.35	.56
	Steps * Group	0.73	.45
Step length	Steps	2.33	.11
	Group	2.12	.15
	Steps * Group	0.17	.83
Step width	Steps	8.94	<.01
	Group	0.01	.96
	Steps * Group	0.25	.69
MoS FW	Steps	4.64	.02
	Group	0.21	.65
	Steps * Group	.792	.43
MoS ML	Steps	1.20	.29
	Group	1.45	.24
	Steps * Group	0.94	.36
Contralateral perturbation First response Paretic Leg			
Step time	Steps	30.6	<.01
	Group	0.15	.69
	Steps * Group	0.37	.63
Step length	Steps	4.89	.02
	Group	0.14	.70
	Steps * Group	.61	.51
Step width	Steps	3.95	.04
	Group	0.11	.73
	Steps * Group	1.48	.23
MoS FW	Steps	.59	.45
	Group	0.08	.77
	Steps * Group	.26	.62
MoS ML	Steps	6.45	<.01
	Group	1.61	.21
	Steps * Group	000	.99
Ipsilateral perturbation First response Non Paretic Leg			
Step time	Steps	24.3	<.01
	Group	**7.34**	**.01**
	Steps * Group	.022	.96
Step length	Steps	27.2	<.01
	Group	3.61	.07
	Steps * Group	.212	.76
Step width	Steps	25.3	<.01
	Group	0.11	.73
	Steps * Group	.03	.96
MoS FW	Steps	28.6	<.01
	Group	3.06	.09
	Steps * Group	0.87	.39
MoS ML	Steps	8.10	<.01
	Group	3.01	.09
	Steps * Group	3.0	.08
Ipsilateral perturbation First response Paretic Leg			
Step time	Steps	10.8	<.01
	Group	**4.35**	**.05**
	Steps * Group	2.84	.07
Step length	Steps	34.9	<.01
	Group	**4.35**	**.04**
	Steps * Group	1.35	.26
Step width	Steps	17.3	<.01
	Group	0.60	.44
	Steps * Group	**5.54**	**<.01**
MoS FW	Steps	19.3	<.01
	Group	3.01	.09
	Steps * Group	**5.98**	**<.01**
MoS ML	Steps	5.26	<.01
	Group	0.13	.71
	Steps * Group	1.41	.25

GG is Greenhouse Geiser correction. MoS is margin of stability. *P*-value for main effect of steps and interaction (Steps * Group) is Greenhouse-Geiser corrected.

for the first step after perturbation, and increased step-widths from the second step onward. MoS values in the ML direction differed from the second step onwards except for the fifth step after gait was perturbed (Fig. 4 contralateral PL and Appendix A). No main effects of group or significant interaction effects with group were found for any of the five gait characteristics, for this perturbation type, see Table 2.

3.2.2. Medio-lateral ipsilateral perturbations

Both ipsilateral gait perturbations, (Fig. 3, ipsilateral) caused a similar change in BoS and step time characteristics for both legs. We found significantly reduced step times in comparison to steady-state step times. Step lengths were reduced for the first two steps and step width increased for all steps after the ipsilateral gait perturbations. When the NPL responded retributions resulted in an increased MoS in ML direction in the first, third and fifth step, moreover FW MoS was reduced in the second step compared to steady-state values (see Fig. 4 ipsilateral NPL and Appendix A). When the PL responded ipsilateral perturbations resulted in a increased MoS in ML direction for the second, fourth and sixth step after gait was perturbed. Furthermore MoS in FW direction was reduced in the second and third step compared to steady-state values (see Fig. 4 ipsilateral PL and Appendix A).

A main effect of group was found for step time when the NPL responded. Post hoc analyses revealed a significant by ($p < .01$) shorter step time in the *F* group in the first step after perturbation. In addition when the PL responded, main effects for group were found for step time and step length (see Table 2). Post-hoc analyses revealed a shorter step time, thus quicker response for the *F* group during the first and second step after perturbation ($p = .03$ and $p = .01$). Moreover step length was reduced in the *F* group during the first step after perturbation ($p < .01$). Furthermore, significant interactions between group and step were found for step width and MoS in FW direction when the PL responded, Table 2. Post-hoc analysis revealed no significant differences between groups in step width, but did reveal a significantly lower MoS in FW direction in the second step in the group of fallers compared to group of non fallers, indicating a reduced dynamic stability ($p < .001$).

Finally for all perturbation types and responding gait characteristics, results were the same when preferred steady-state gait speed was included covariate, see Appendix B.

3.2.3. Anterior-posterior decelerating gait perturbations

After gait was perturbed with a deceleration of the split belt, (Fig. 5) independent from which leg responded, the first step response was a shorter step (both in terms of time and length). Moreover step width was increased for all consecutive steps after the perturbation. MoS did not differ compared to steady-state values when the NPL responded. MoS in the ML direction increased for the first and second step if the PL responded and MoS in FW was reduced in the third step (Fig. 6 decelerating PL and Appendix A). No main effect of group was found for neither responding leg. Two significant interaction effects between steps and group on

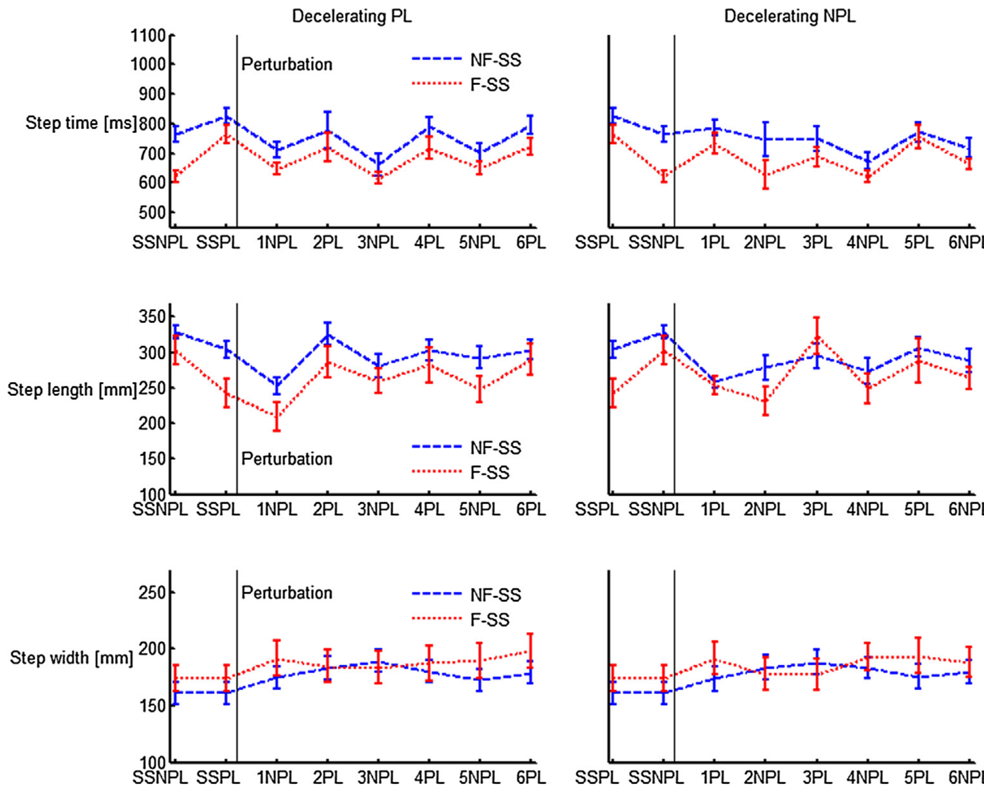

Fig. 5. Step time and base of support (BoS) gait characteristics during steady state (SS) and after gait was anterior-posterior perturbed for the paretic leg (PL) and non paretic leg (NPL).

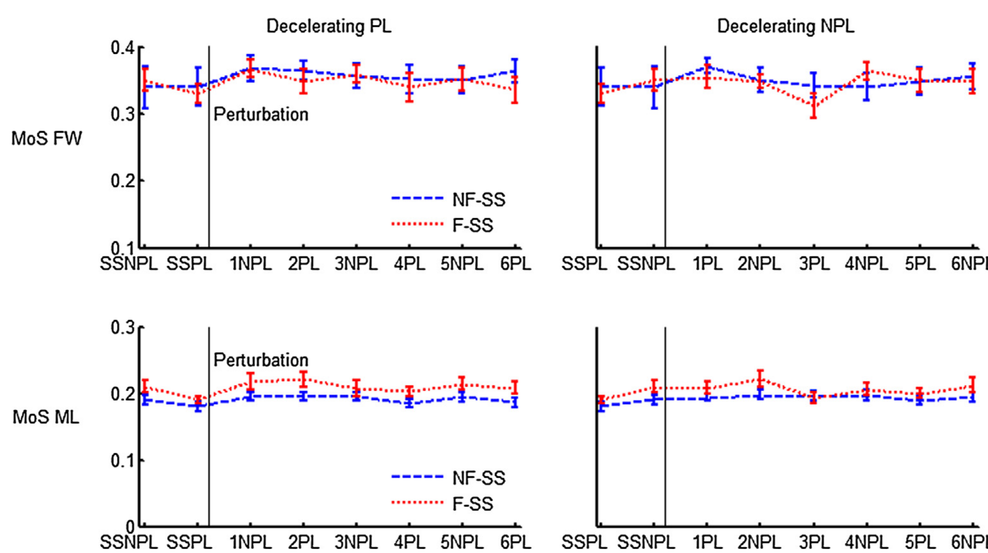

Fig. 6. Margins of Stability (MoS) in the forward (FW) and medio-lateral (ML) direction during steady state (SS) and after gait was anterior-posterior perturbed for the paretic leg (PL) and non paretic leg (NPL).

step width were found for both perturbation types (see Table 3). However post hoc analysis revealed no differences in step widths between groups.

4. Discussion

Our aim was to explore whether differences exist in responses to external gait perturbations between a group of stroke survivors that experienced a fall in daily life, and a group that did not. The gait perturbations resulted in significant deviations in gait characteristics, which indicates that gait adjustments were made. We found that both groups of stroke survivors react largely similar to the gait perturbations. More specifically, the strategy of reacting with longer/shorter steps to certain gait perturbations was similar, as step times did not differ between groups. In addition, those responses were similar to what we expected for ML perturbations illustrated in Fig. 1. Furthermore, BoS characteristics showed similar decreasing trends over consecutive steps between groups.

Table 3
Mixed model ANOVA for AP gait perturbations. With the gait characteristic as dependent variable. Number of steps as within factor and group as between effect. Significant group and interaction effects are printed in bold.

Gait characteristic	Effect	*F*	*P*-value
Decelerating FW perturbation First response Non Paretic Leg			
Step time	Steps	0.86	.40
	Group	0.77	.39
	Steps ∗ Group	.026	.94
Step length	Steps	5.04	.02
	Group	.08	.79
	Steps ∗ Group	1.91	.17
Step width	Steps	.383	.61
	Group	.001	.99
	Steps ∗ Group	**3.88**	**.04**
MoS FW	Steps	4.81	.01
	Group	.406	.53
	Steps ∗ Group	.704	.49
MoS ML	Steps	2.60	.10
	Group	1.02	.33
	Steps ∗ Group	2.07	.15
Decelerating FW perturbation First response Paretic Leg			
Step time	Steps	6.18	.01
	Group	0.48	.49
	Steps ∗ Group	.26	.69
Step length	Steps	8.53	<.01
	Group	2.01	.17
	Steps ∗ Group	.08	.86
Step width	Steps	.55	.55
	Group	.06	.80
	Steps ∗ Group	**6.15**	**<.01**
MoS FW	Steps	.75	.44
	Group	.02	.88
	Steps ∗ Group	.48	.55
MoS ML	Steps	.47	.55
	Group	2.73	.11
	Steps ∗ Group	.48	.54

GG is Greenhouse Geiser correction. MoS is margin of stability. *P*-value for main effect of steps and interaction (Steps ∗ Group) is Greenhouse-Geiser corrected.

However, for ipsilateral ML perturbations the *F* group reacted quicker and with a reduced step length in the first step. Nevertheless, MoS values between groups were similar and MoS values did not deviate from steady-state MoS values (Table 2 and Figs. 4 and 6). Therefore, it seems that both groups of stroke survivors were able to adequately respond to the gait perturbations. However, after gait was perturbed with an ipsilateral perturbation and the paretic leg (PL) responded fallers showed a significantly lower MoS in FW direction during the second step, suggesting lower stability. This is somewhat puzzling, because this perturbation disturbs gait in the ML direction. Possibly, widening the step while maintaining FW MoS when stepping with the paretic leg was challenging for this group.

To better understand this finding, we extended our analysis by studying the velocity of the center of mass in FW direction and the trunk angle for this particular gait perturbation in the FW direction. While the fallers group were able to increase their step width sufficiently and thereby restoring ML MoS, this came at the expense of a reduced step length, due to constant treadmill speed. This led to a more rearward position on the treadmill. To compensate for this change in position on the treadmill, fallers group attempt to regain speed by creating a larger forward momentum by a more forward shifted trunk during the second step, which then led to a smaller FW MoS. Although MoS in FW direction was decreased in the *F* group it may not be representative for everyday life situations where we would expect that one would try to slow down or even stop during the second step rather than trying to speed up. Thus, gait characteristic responses from the second step onward when the perturbations are applied on a treadmill with a constant belt speed may not be representative for real-life situations.

At present, only a few studies have applied larger external gait perturbations in stroke survivors (Kajrolkar et al., 2014; Kajrolkar and Bhatt, 2016; Krasovsky et al., 2013). While Krasovsky et al. (2013) found a larger global response in terms of strategy and timing of gait rhythm after gait was perturbed in stroke survivors compared to healthy older adults (Krasovsky et al., 2013). Kajrolkar et al. (2014) concluded that stroke survivors have a preserved ability to adjust gait characteristics and maintain dynamic stability (Kajrolkar et al., 2014). Our AP decelerating perturbations tended to cause a backward fall, however, contrary to the studies of Kajrolkar et al. (2014) and Kajrolkar and Bhatt (2016) our participants did not make a backward step, instead all participants were able to continue to move forward. It is interesting to see that apparently small differences in onset and magnitude of the perturbation can result in such different responses.

Our study is not comparable to any previous study executed in stroke survivors, since to the best of our knowledge this was the first study assessing differences in responses to larger external gait perturbations between fallers and non fallers in stroke. Our results indicate that perturbation responses are not useful as predictors of fall risk, which is different from perturbations during standing (Mansfield et al., 2015b). This suggests that priority should be given the study of steady-state gait characteristics in stroke survivors are more promising regarding predicting fall risk (Mansfield et al., 2015b; Punt et al., 2016). Nevertheless, gait perturbations might be useful in fall prevention programs, as perturbation based gait training appears to be effective in fall prevention in older adults and in people with Parkinson's disease (Mansfield et al., 2015a).

There is a number of possibilities that might explain our limited findings. First, perturbations applied might lack ecological validity. Second, the perturbation magnitude may have been too small. MoS in the first step after gait perturbations were equal or even slightly increased in comparison to steady-state values, which may indicate that the perturbation magnitude was not challenging enough to differentiate between groups. Each perturbation type was repeated four (ML perturbations) and eight (decelerating perturbations) times, however due to handrail grasping we analyzed only the final two perturbations and thereby gathering the average response. From a different perspective, we may argue that perhaps only the response to the first gait perturbation is relevant for fall risk, as during a perturbation in daily life, people have only one chance to respond adequately and thereby prevent an actual fall incidence. Finally, it may be that small differences between groups are present, yet not found in this study due to the limited sample size.

Another methodological consideration is the gait speed during the perturbations. We used a fixed speed thereby making sure that the applied perturbations were similar across participants. Changing the treadmill speed to somebody's preferred speed means that the applied perturbation is executed over another percentage of the gait cycle as the duration of the gait cycle will change with speed while the duration of the ML displacement does not. Adjusting gait speeds would thus actually result in different gait perturbations, which makes it unfair to compare between participants. However, perturbing gait at preferred speed is more ecologically valid, since most perturbations experienced during gait in daily life will occur at preferred speed. Nevertheless, in this case it would remain unclear whether differences between groups would be due to how they respond or due the fact that perturbations were different. However, given the problems associated with designing "matched" perturbations at subjects preferred speeds, we choose to perturb subjects at a fixed speed. Finally our sample of stroke survivors may not be representative of the entire population based on the ratio male/female participants.

In conclusion, this study found limited differences in gait perturbation responses between stroke survivors that fell and that did not fall during follow-up. Although step length after an ipsilateral perturbation when the paretic leg responded was reduced in our group of fallers, this did not result in smaller MoS values than in non-fallers. Furthermore the FW MoS during the second step after a medio-lateral ipsilateral gait perturbation where the paretic leg responded differed between fallers and non-fallers, but this was most likely not directly caused by the perturbation itself but rather by the need to keep up with the belt speed. Our results do not support the use of gait characteristic responses to predict fall risk. However, our sample size was limited, and a larger cohort might reveal differences which were not found in the present study.

Conflict of interest

Sanne Roeles worked at Motekforce Link B.V., The Netherlands who had no influence in the execution, analysis or writing of this manuscript.

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Appendix A. Supplementary material

Supplementary data associated with this article can be found, in the online version, at http://dx.doi.org/10.1016/j.jbiomech.2017.02.010.

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